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IN THE SPECIFICATION

Please amend p. 1, [0003] as follows:

[0003] MR elastography (MRE) involves measuring motion resulting from low frequency vibration. Present MR elastography methods use a separate gradient waveform to encode the motion, for example, in <u>the</u> context of the Larmor equation that is used to measure tissue strain and discussed in United States Patent No. ~~5,982,828~~ <u>5,952,828</u> issued to Rossman et al. The gradient waveform may be added between the RF excitation and the readout of the echo. The resulting increased echo time has the undesirable effect of decreasing the signal amplitude, as well as increasing the imaging time.

Please amend p. 4, [0012] as follows:

[0012] The present system addresses the above and other problems, thereby advancing the state of the useful arts, by utilizing one or more imaging gradient waveforms to identify motion of the specimen in an MRE system. All of the gradients accumulate phase from motion and, consequently, may be used to encode the phase. Echo times are advantageously reduced because the imaging gradients encode a harmonic motion, as opposed to separate motion-encoding gradients. Some distortion is possible in this approach. However, the distortion of the phase caused by simultaneous<u>ly</u> encoding position and motion is minimal because the phase accumulated by the motion is small compared to the phase changes generated during frequency encoding. Imaging times for 100 Hz vibrations can be reduced by a factor of from 2 to 3 or more. Moreover, the echo times can be reduced by a factor of more than three which increases the signal to noise ratios (SNR) significantly in a positive way. There is a reduction in the sensitivity to motion so larger amplitude motions are desired. However, the increase in SNR compensates to some extent for the loss in sensitivity, even with no increase in the amplitude of the motion.

Please amend p. 5, [0014] as follows:

[0014] Microwave radiation includes a band ranging from 300 <u>MHz</u>~~Mhz~~ to 30 <u>GHz</u>~~Ghz~~. These frequencies are generally useful for both MRMA and MRMT.

Please amend p. 9, [0034] as follows:

[0034] The microwave power generator 122 (see FIG. 1) coupled with the microwave amplifier 124 having, for example, 2 kW power and 0.5-1 μsec. pulse width provides sufficient degrees-of-freedom to drive the microwave illumination. Sources with higher power and broader band can also be used, but they tend to be more expensive. The power signal source and its associated parameters (repetition rate and pulse width) are broadly selectable because the thermoelastic response is not critically dependent on these parameters. While the results using high frequency (3 GHz and 9 GHz) in Wang *et al.* are very impressive, lower <u>frequencies</u>~~freuqencies~~ (300-450 MHz) are preferred because they provide adequate depth penetration without over-exposing superficial tissues to microwave radiation. Another reason for using lower frequencies is that when MRMA is combined with MRMT, when using glycerin as the coupling fluid 142, the use of higher frequencies above 500 MHz results in higher electromagnetic loss to the fluid 142. Alternatively, the clinical interface 108 may be implemented using phased arrays, such as a 4-quadrant array with independent power switching for heating purposes to provide the advantage of partial breast illumination. Such a design is even more <u>preferable</u>~~preferrable~~ when MRMT is integrated with MRMA because it enables easier separation of the field effects which need to be normalized relative to the responses from other portions of the breast when simultaneous illumination is not applied.

Please amend p. 10, [0039] as follows:

[0039] Increased image encoding efficiency becomes more important in multispectral imaging applications. It is excessively time-consuming to measure multiple motion frequencies with spin echo pulse sequences. The pulse sequence can be improved by multiplexing a number of encodement types obtained from each motion-encoded signal. For example, fast spin echo, echo planar and/or[[or]] spiral scanning methods can be used to image a plurality of motion frequencies by assigning an encodement type to a particular frequency or frequency band. Moreover, the frequency of the motion-encoding gradients can be changed, and the sequence of microwave pulses can also be changed relative to the motion-encoding gradient to alter the frequency response of the motion.

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Please amend pp. 11-12, [0041]-[0043] as follows:

[0041] The system 100 (see Figs. 1 and 2) also provides for methods of performing microwave tomography in an MR environment. The system elements used for MRMT include a monopole antenna array 114, the motion mechanism 154 for multi-slice imaging and the coupling fluid 142. One predominant consideration of the system design includes space constraint that is imposed by where the system will reside in the intended environment~~envoronment~~ of use, namely, the bore of an MRI machine, such that the system 100 will operate according to design parameters without interfering with the magnetic~~magentic~~ resonance function.

[0042] FIG. 3 shows a structure of a monopole antenna element 160, which may contain a central conductor 161 surrounded by a teflon insulator 162, an upper exposed dielectric support~~suport~~ section 163, and a lower dielectric support 164. Monopole antenna performance, as measured in terms of characteristic impedance and radiation patterns, can be achieved using an array that contains a plurality of monopole antenna elements 160, the elements 160 presenting a range of differing diameters. A design parameter for achieving repeatable performance is that each of the elements 160 have a characteristic impedance, for example, 50 Ω. The design of monopole antenna elements 160 may be modified to improve performance in an MR environment by reducing the size of metallic structures, such as central conductor 161, to reduce or eliminate MR signal artifacts. As shown in FIG. 3, the current monopole antenna design is scaled down to an outer diameter D of 2.2 mm, and non-metallic dielectric materials are used in the upper dielectric support 163 and lower dielectric support 164.

[0043] Microwave tomography of the breast is facilitated by a low-contrast (as compared with the tissue) and moderately lossy coupling fluid 142 which enhances broadband operation of the transceiving antenna array and diminishes surface reflections from the breast-fluid interface that may distort image quality, especially with respect to the resolution of internal breast structure. The imaging chamber 140 is constructed to hold the coupling fluid 142 into which the antenna array 114 and the "compression" plate 128 are inserted. The coupling fluid 142 is preferably a 70-90% glycerin solution (70-90% glycerin with 10-30% water by volume). Glycerin mixed with water has advantageous spectral properties that ~~that~~ do

not cause high electromagnetic loss at high frequencies (700 MHz to 3 GHz) within the imaging chamber 140. This low attenuation is especially useful for high frequency broadband monopole antenna operation, and signal attenuation is further reduced at lower frequencies below 500 MHz. The reduced signal losses observed through glycerin and water frequencies below 500 MHz comprises an improvement relative to other possible coupling fluids and is advantageous for the MRMA system.

Please amend p. 15, [0049] as follows:

[0049] MRMA image reconstruction may be implemented by first including the microwave absorption as an unknown source term and estimating its spatially-dependent value as an additional inversion parameter. This process can be implemented in several steps. Initially, the tissue elastic properties are assumed to be known and the driving source of mechanical motion is the only parameter that needs to be estimated. In another embodiment, magnetic resonance elastography data can also be collected immediately prior to MRMA excitation which can supply the spatial map of tissue elastic properties for use in MRMA image reconstruction. The algorithm can be further expanded to include mechanical property estimation. The second algorithm accepts anatomical priors derived from the MR magnitude image volume (or additional image acquisition sequence) using advanced tissue segmentation and mesh generation capabilities. The estimation of mechanical~~mechnical~~ properties can be constrained to adhere to the prior segmentation, while tissue elasticity ranges can be derived from literature values for fat and fibroglandular tissue. These elasticity constraints coupled to full freedom in terms of estimating the local power deposition make it necessary for the algorithm to emphasize the latter during data-model minimization. Maps of microwave absorption can thus be reconstructed based on this algorithm. While such maps are likely informative, their diagnostic value may be limited because the power absorption map also encodes extrinsic factors unrelated to tissue pathology. By way of example, any mapping technique, such as a multivariate correlation or neural network, may be used to relate elastic~~eleastic~~ properties reconstructed directly from MRMA data to conductivity values used by the MRMT algorithm.

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Please amend p. 17, [0055] as follows:

[0055] Simulations of the reconstruction of position and phase using a standard frequency encoding gradient during motion like that induced in MR elastography reveal that the magnitude and phase of the reconstructed images are perturbed by the motion especially around sharp edges but that ~~that~~ perturbation is relatively small.

Please amend pp. 17-18, [0059] as follows:

[0059] The simple case is finding motion in just the frequency encoding direction. FIGS. 8A and 8B show a pulse sequence 800 and reverse pulse sequence 810 having ~~with~~ the sign of the frequency encoding gradients 804 and 814 reversed with respect to each other. Note that 804 and 814 are the frequency encoding gradients. The images can be combined by electronically flipping the images obtained from the second pulse sequence in the frequency encoding direction so the positions are aligned. Then the phase of each pixel is subtracted and the result is the phase accumulated from the motion in the frequency encoding direction. The echo time is reduced substantially by removing the motion encoding gradients and the total imaging time is also reduced. Because the signal decays exponentially, the signal from the pulse sequences in FIGS. 8A and B is much greater than that from FIGS. 7a and 7B. Pairs of images must still be obtained at several relative phases between the harmonic motion and the imaging gradients.